Development of a Standing-Up Motion Guidance System Using an Inertial Sensor

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Abstract. The standing-up motion consists of (1) a flexion phase, in which the center of gravity (COG) moves forward, and (2) an extension phase, in which COG raises upward. However, because it is difficult for elderly and disabled people to combine both phases, they need to perform each phase individually. Although most people who are unable to stand up are able to raise their COG upward, they are unable to move it forward. Therefore, we proposed a system and evaluated its efficacy in supporting forward COG movement.

Keywords: standing-up motion, center of gravity, inertial sensor.

1 Introduction

Standing up from a chair is a very important motion in daily life. The standing-up motion is complicated because it involves a change in the center of gravity (COG) from over the ischium to over the feet. If a person cannot transfer their COG onto their feet, standing up from a chair becomes difficult because the muscles required for that motion are not activated. The standing-up motion is considered to consist of (1) a flexion phase, in which the COG moves forward as the trunk leans forward, and (2) an extension phase, in which COG raises upward as the trunk lifts. Generally, a healthy person skillfully uses a combination of these two phases to stand up from a chair. However, it is difficult for elderly and disabled people to combine these phases; therefore, they need to perform each phase individually. In addition, most people who cannot stand up from a chair are able to raise their COG upward, but are unable to move their COG forward and complete the standing motion. These people need assistance during the flexion phase rather than the extension phase. In medical institutions, caregivers provide assistance by pushing the patient's trunk forward until the person can raise their COG upward. To provide similar assistance, we would like to develop a system to support forward movement of the COG during the standing-up motion.

To realize such a system, both the standing-up motion and the COG position need to be measured. This is easily carried out using a force plate and a three-dimensional motion capture system. However, because the measurement equipment is large and expensive, daily usage of such a system can be strenuous [1]. In order to obtain appropriate measurements without using complicated equipment, we hypothesized that

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trunk movement needed to be measured in the flexion phase so that COG movements could be estimated using only the trunk movement data. This hypothesis is based on the fact that only the trunk moves in the flexion phase of standing up from a chair. Therefore, we acquired the trunk angle using an inertial sensor installed on the trunk and calculated the COG movement in real time during standing by applying the trunk angle to a human body model. If a patient can sense when their COG has been transferred to their feet, standing up from a chair would become easier because after being alerted of the COG transfer, the patient would only need to perform the upward motion. Fig. 1 shows an illustration of our system in which the inertial sensor data is wirelessly transmitted to a computer that notifies the subject of the COG transfer using visual and auditory signals. This study describes the optimal position at which the inertial sensor should be set on the human body model to allow estimation of the COG position. Furthermore, we report the efficacy of this COG estimation method.

Fig. 1. Depiction of the COG estimation system

2 Optimal Position of the Inertial Sensor

2.1 Experimental Setup and Procedure

We used two different human body models in this study, the three-link model and the four-link model, shown on the left side of Fig. 2. Both models define the trunk angle differently: In the three-link model, the trunk angle was defined as the link that connected the acromion–trochanter major, whereas the angle in the four-link model, the trunk angle was the link that connected the acromion–ilium. Furthermore, to determine the most appropriate model for estimating the horizontal COG position during the flexion phase and the optimal position of the inertial sensor on a trunk, two inertial sensors were attached to the trunk: one on the sternum and the other on the ilium. In addition, we placed three infrared reflection markers on the acromion, ilium, and trochanter major, (right panel of Fig. 2). Subsequently, we measured body movement using the three-dimensional motion capture system (Detect Inc.) with four infrared cameras sampling at 60 Hz. The measurement specifications of the inertial sensor (Logical Product Inc.) were 300 °/s for the gyroscope and 5 g for the accelerometer. Three healthy male subjects were asked to bend their trunks under two different conditions: (1) straightening the back and (2) hunching the back, as shown in Fig. 3. These measurements were repeated two times for each subject.

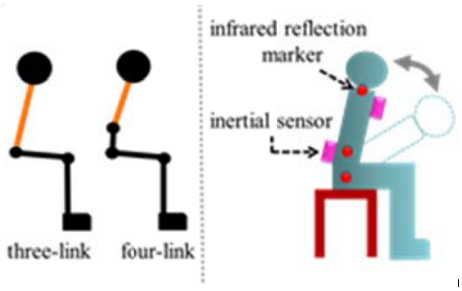

Fig. 2. Three-link and four-link models (left) and the experiment schematic view (right)

Fig. 3. Straightening the back (left) and hunching the back (right)

2.2 Experimental Results

Fig. 4 shows an example of the trunk angle results we obtained using the three-link and four-link models, which were calculated from the motion capture data. From all of the results, the root mean square (RMS) between the two models was 3.10°. This value was considered to be small because measurement errors of a few degrees may have occurred because of the infrared markers moving slightly on the skin [2]. On the basis of these results, we decided to use the three-link model because it was easier to operate. To determine the most suitable position for the inertial sensor between the sternum and ilium, we calculated the RMS error (RMSE) values of the trunk angles. The RMSE values quantify the differences between the trunk angle obtained from the inertial sensor and those obtained from the motion capture system. The results are shown in Table 1. We observed that when the inertial sensor was placed on the ilium, the resulting trunk angle was significantly influenced by trunk motion (RMSE of 6.7 with a straight trunk increased to 13.67 with a hunched trunk). This was because the ilium did not move enough while the back was hunched; therefore, the sensor located on the ilium could not accurately measure the trunk angle. However, when the sensor was placed on the sternum, the trunk angle did not change significantly with back hunching. Therefore, we concluded that attaching the inertial sensor to the sternum was optimal for this experimental setup.

Table 1. RMSE values of the trunk angle for each sensor position when two different trunk motions were performed

	straightening the back	hunching the back
sternum	5.87	7.84
ilium	6.7	13.67

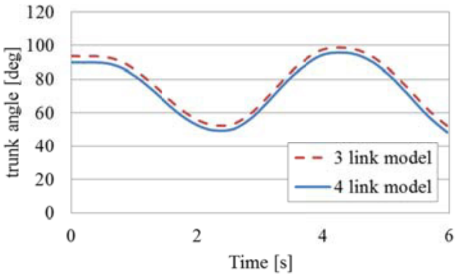

Fig. 4. Trunk angle measurements from the three-link and four-link models, which were calculated from the motion capture data

3 Estimation of Horizontal COG Position Using Trunk Movement Measurements

3.1 Estimation of the Horizontal COG Position

This section describes the method for determining the horizontal COG position from the trunk angle data acquired from the inertial sensor. As shown in Fig. 5, we assumed the human body to be a rigid body with three links. The loads on the foot (R1) and the chair (R2) were estimated using an equation of motion, as shown in Fig. 5. In addition, the horizontal COG position was estimated from the moment of force among P1, P2, P3, R1 and R2. The rigid body model was newly made based on the results of other studies [3, 4].

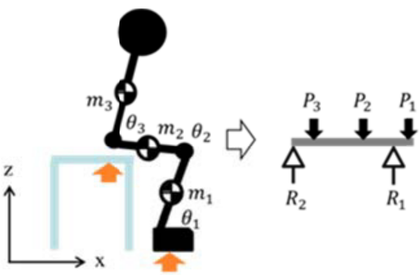

Fig. 5. The human body model and the method used to calculate the horizontal COG position

3.2 Evaluation of the Estimation Method for Horizontal COG Position

We compared the horizontal COG position that we estimated from the inertial sensor data with that from the three-dimensional motion capture data and force plate data. First, the standing-up motion was measured by combining the three-dimensional motion capture system (Vicon-Peak Inc.) with eight infrared cameras and four force plates (AMTI Inc.). Twelve infrared markers were placed on both sides of the acromion, iliac crest, greater trochanter, knee, ankle, and metatarsal bone. The subjects were five non-disabled males; they were asked to stand up from a chair at two different speeds: (1) at their usual standing speed and (2) at a speed that was slower than usual, resembling that used by elderly people. The height of the chair was 0.42 m, and the subjects had to decide upon a suitable foot position upon standing. Measurements were repeated five times for each subject and each condition. Data were recorded using cameras sampling at100 Hz.

Second, the motion capture system data were used to calculate the trunk angle and the horizontal COG position was estimated using the trunk angle in our method. We decided that the starting point of the horizontal COG would be the center of the foot heel, and forward positions were given positive values.

Third, these estimated horizontal COG positions were compared with those obtained after combining the motion capture system and force plate data.

Examples of results are shown in Fig. 6. The left graph of Fig. 6 shows how horizontal COG position changed when the subject stood up with their usual speed, and the right graph of Fig. 6 shows the changes in COG at a slower standing speed. The vertical axis represents the horizontal COG position in meters and the horizontal axis shows the time in seconds. The solid and dotted curves show the measured and estimated data, respectively. The vertical dotted line shows the time at which the buttocks left the chair and the horizontal dotted line represents the boundary at which the base of support on foot was located. This boundary of support was defined to be −0.053 ± 0.0057 m from the heel [4].

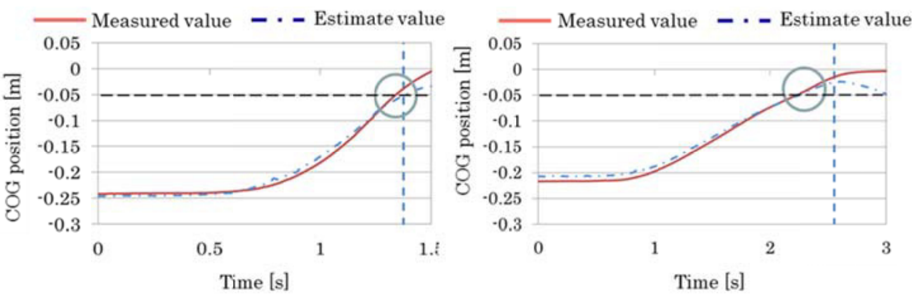

Fig. 6. Standing up with a usual speed (left) and a slower-than-usual speed (right)

The RMSE values for estimated and calculated horizontal COG positions are shown in Table 2. We found that the mean RMSE values associated with standing up

at a normal speed and standing up slowly were both approximately 0.02 m. As a result, we determined that our method estimates horizontal COG position with relatively good accuracy because there were no major differences between both conditions.

We found that differences between the estimated and measured data became larger once the vertical dotted line in Fig. 6 was crossed. In this experiment, the time when the buttocks left the chair was defined as the time at which the value of force plate (placed underneath the chair) became zero. However, our method was valid while the buttocks remained in contact with the chair, but it resulted in larger differences after the buttocks left chair.

In our system, it was important to estimate the horizontal COG position immediately before it reached the base of support boundary (horizontal dotted line in Fig. 6) in order to judge whether the horizontal COG position was within the base of support. From Fig. 6b, we found that when the subject stood up slowly, the buttocks left the chair after the horizontal COG position had completely entered the base of support. Therefore, we concluded that our system can estimate horizontal COG position until the horizontal COG position entered the base of foot support because the buttocks did not leave when the subjects stood up slowly.

Table 2. RMSE values for the horizontal COG position in meters

Subject	Normal Standing-up	Slow Standing-up
A	0.014	0.008
B	0.042	0.044
C	0.018	0.018
D	0.019	0.015
E	0.010	0.010
Mean	0.021±0.011	0.019±0.013

4 Proposition of the Standing-Up Motion Guidance System and Evaluating Its Efficacy

4.1 Proposing a Standing-Up Motion Guidance System

Considering the results of sections 2 and 3, we manufactured a trial version of the standing-up motion guidance system. Our system comprised an inertial sensor and a computer. The measurement specifications of the inertial sensor (Logical Product Inc.) were 300°/s and 5 g. A software program, which was developed in Visual C#, was used to estimate the horizontal COG position and judge whether the horizontal COG position was inside the base of support. Then, the time elapsed as the trunk raised upward was monitored through an instruction display, shown in Fig. 7.

Fig. 7. Display screen for the trial version of the guidance system

After the inertial sensor was placed on the user's chest, the number of bars in Fig. 7 increased as the user leaned his/her trunk forward. In this trial version, red bars and the instruction "Lean the trunk" were displayed when the horizontal COG position was between 0.06 m and 0.12 m from the boundary of the base of support. When the COG position was between 0 and 0.06 m, the interface displayed yellow bars and the instruction "Lean a little more." Once the COG position entered within the base of support, blue bars were displayed and the interface gave the instruction "Raise your trunk" along with an auditory beep.

4.2 Evaluation of our Guidance System

We used data from an electromyogram (EMG) to evaluate the efficacy of our guidance system. Four experimental patterns were prepared: (1) Subjects were asked to stand up slowly when the system judged that their horizontal COG position had not entered the base of support; (2) Subjects were asked to stand up slowly when the system judged that their horizontal COG position was very close to the boundary of the base of support; (3) Subjects were asked to stand up slowly when the system judged that their horizontal COG position was inside the base of support; and (4) Subjects were asked to stand up without using the guidance system. Two non-disabled males participated in this experiment and repeated all conditions five times.

During the experiment, EMG data from muscles related to the standing-up motion were obtained. Data from the tibialis anterior (TA), rectus femoris (RF), gluteus maximus (GMA), and erector spinae (ES) were measured using an electromyogram (Medicament Inc.). However, only muscles on the right side were measured because the standing-up motion is generally symmetrical about the median sagittal plane. Before experimentation, the maximum voluntary contraction (MVC) EMG data for all muscles were measured, and the EMG data during the standing-up motion were normalized by the EMG of MVC. We defined this normalized data as %MVC.

Results are shown in Figs. 8 and 9. The vertical axes represent %MVC and horizontal axes represent the experimental pattern. For pattern (1), the subjects were not all successful at standing up during all trials. Generally, by comparing patterns (2)–(4), the %MVC values from pattern (2) were relatively larger than those from patterns

(3) and (4). Moreover, we did not find any large differences between patterns (3) and (4). Therefore, we considered that our system was able to guide subjects to the optimal forward trunk position that would allow them to stand up easily.

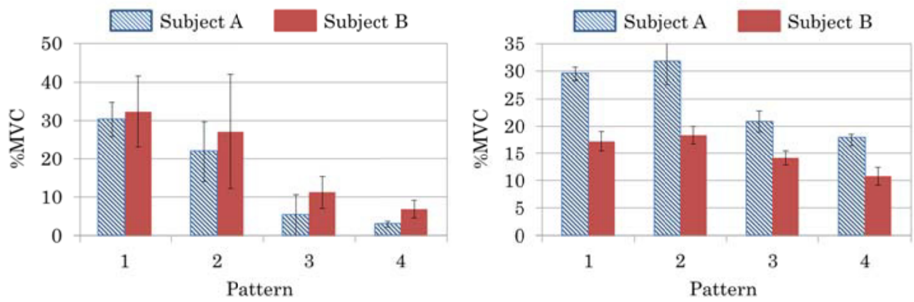

Fig. 8. %MVC results for TA (left) and RF (right)

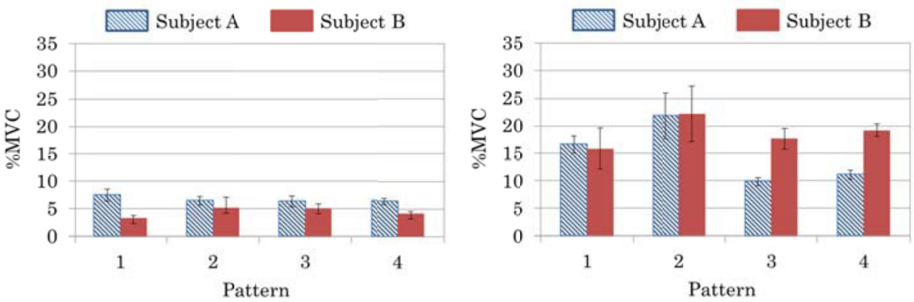

Fig. 9. %MVC results for GMA (left) and ES (right)

5 Conclusion

In this paper, we proposed and evaluated the efficiency of a standing-up motion guidance system that informs the user of the optimal time at which their trunk should be raised when standing up from a chair. However, there are many problems that remain unresolved. In subsequent research, we plan to perform the following activities:

(1) Produce a method to estimate three-dimensional COG position
(2) Improve the user interface
(3) Provide significant evidence through more subjects and perform more trials
(4) Evaluate our system using patients who train standing-up motion in the clinical site

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